The Effect Of Vibratory Noise Input On Postural Responses To An Unexpected Loss Of Balance

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Dedication

For my mom and dad.

Abstract

Vibratory noise input applied to the foot sole has been shown to improve static balance control across various populations (i.e., younger adults, older adults, individuals with diabetic neuropathy etc.). However, there is little research on whether vibratory noise improves reactive balance control. This is concerning because falls typically occur when an individual is unable to quickly recover from a loss to their balance. Therefore, the purpose of this thesis was to examine whether vibratory noise input affects postural responses following an unexpected surface translation. A secondary aim was to examine the effects of vibration on static balance performance to replicate previous findings. Eighteen adults (10 females and 8 males) completed six quiet standing trials and 28 surface translation trials. For all trials, participants stood barefoot, while blindfolded and wearing headphones. Three vibrating elements were placed directly underneath each foot (i.e., one each at the first metatarsal, fifth metatarsal and at the heel). For each standing trial, participants were instructed to stand quietly. For each surface translation trial, participants were instructed to recover their balance without stepping in response to a unexpected surface translation. Participants were unaware of which trials did or did not have vibration applied to the foot soles. Static and reactive balance control were quantified using various kinematic, kinetic and electromyography (EMG) measures, while the ability to recover balance was quantified through the measurement of EMG and body kinematics. Results indicated that vibratory noise input did not influence most measures of static and reactive balance control. This suggests that the application of vibratory noise input to the foot soles is not beneficial in younger adults. Future studies should replicate this study with clinical populations to determine whether the benefits of vibratory noise input are limited to individuals with worsened balance ability.

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Table of Contents

List of Figures

backward direction, and (d) the Soleus muscle in the backward direction between vibration conditions. TO= Translation Onset, MO= Muscle Onset.

List of Tables

Abbreviations

AP = Anterior-Posterior

BOS = Base of Support

CNS = Central Nervous System

COM = Centre of Mass

COP = Centre of Pressure

EMG = Electromyography

MO = Muscle Onset

ML = Medio-Lateral

MVC = Maximum Voluntary Contraction

NV = No Vibration

RMS = Root Mean Square

SD = Standard Deviation

SOL = Soleus

TA = Tibialis Anterior

TO = Translation Onset

V = Vibration

1. Reactive balance control: Introduction

The incidence of falls is becoming increasingly common across the lifespan. Additionally, injurious falls have become a significant concern for healthcare costs, with falls being the third leading cause of unintentional injuries in younger adults (Heijnen & Rietdyk, 2016) and the third leading cause of death across all ages (Heijnen & Rietdyk, 2016). Since falls are most often the result of a trip (i.e., 48%-60% of falls) or a slip (i.e., 25% of falls in younger adults and 50%-60% of falls in older adults) (Berg et al., 1997; Heijnen & Rietdyk, 2016; Horak et al., 1997; Pavol et al., 2001), it is important to understand how individuals typically respond to an unexpected loss of balance and investigate how different neuromuscular factors might contribute to one's ability to recover balance.

1.1. Reactive balance control: Postural strategies

One method used to study balance recovery mechanisms is to administer a support surface translation. This involves rapidly moving the surface on which the person is standing to rapidly perturb the individual's centre of mass (COM). Using this method, researchers have observed different balance recovery strategies that occur in response to the unexpected postural disturbance. Some individuals rely on a fixed-support strategy (i.e., ankle, hip or a combined strategy) that restores the COM without having to change the base of support BOS. When an ankle strategy is employed in response to a forward or backward translation, the body moves about the ankles, and the ankle muscles are activated to allow the centre of pressure (COP) to move past the COM to restore it within its BOS (Horak et al., 1997; Horak, 2006; D. Winter, 1995). For perturbations with a greater acceleration or those occurring in the medio-lateral direction, other muscles (e.g., hip and trunk muscles) will need to be activated to restore the COM within its BOS. Lastly, a combined strategy, where the body moves about the ankles and hips, could be employed to maintain balance when perturbed (Horak et al., 1997). When a fixed-support strategy is inadequate to recover balance due to too large of a perturbation or other task constraints, a change-in-support strategy that restores the COM by changing the BOS may be used instead (McIlroy & Maki, 1993a). This involves individuals stepping to widen their BOS or grasping onto a different support surface to change their BOS (Maki et al., 1999, 2008; Maki & McIlroy, 2006). Although change-in-support strategies are typically employed when perturbations are so great that the BOS has to be changed in order for balance to be successfully restored, individuals may also preferentially adopt a change-in-support strategy in response to smaller perturbations (McIlroy & Maki, 1993b).

1.2. Reactive balance: The effects of aging and sensory neuropathies on balance recovery

The ability to recover balance from an unexpected loss of balance is deteriorated in older adults and adults with sensory disorders (e.g., neuropathies) and consequently, they experience an increased risk of falling (Timar et al., 2016). Older adults and adults with neuropathies have been observed to have delayed and diminished balance correcting responses. Compared to younger adults, older adults demonstrate a 10-15 ms delay in lower limb muscle activation following an unexpected surface translation (Lin & Woollacott, 2002; Tokuno et al., 2010). Older adults also activate their proximal and distal muscles to a greater extent as well as exhibit more mediolateral instability (Allum et al., 2002; Lee et al., 2018). The delayed and increased muscle activation patterns may explain why compared to younger adults, older adults are more reliant on a hip strategy than an ankle strategy (Tokuno et al., 2010).

Similar to older adults, adults with sensory neuropathy responding to a surface translation have a 20-30 ms delayed muscular activation and up to a 50% reduced postural response magnitude scaling compared to age matched controls (Dickstein, 2003; Inglis et al., 1994).

1.3. Reactive balance control: The role of the somatosensory system for balance recovery

One reason for the decreased balance recovery responses in both the older adult and sensory neuropathy populations may be a decline in somatosensation. The somatosensory system is essential for reactive balance control because it provides proprioceptive (e.g., information about the position and length of muscles and joints) and tactile information (e.g., information about pressures and ground reaction forces against the feet) (Kars et al., 2009). When one's support surface is perturbed, feedback from the somatosensory system helps to scale changes in the COP to allow the central nervous system (CNS) to provide the appropriate strategy to restore balance. Thus, it makes sense that when there is decreased somatosensation, as has been observed in older adults and adults with sensory disorders, there will be an inefficiency of postural responses. While many sources of proprioceptive and tactile information are likely to contribute to reactive balance control, the focus of this thesis is on the role of the cutaneous mechanoreceptors. Before discussing how previous interventions have aimed to alter tactile sensation to improve balance stability, it is important to first understand the role of the cutaneous mechanoreceptors for balance recovery.

1.4. The tactile system

The tactile system consists of a group of free nerve endings of sensory afferents located beneath the skin known as cutaneous mechanoreceptors (Kennedy & Inglis, 2002; Mildren et al., 2016; Strzalkowski et al., 2018). These mechanoreceptors, especially those within the glabrous skin of the feet, provide sensory information on changes in pressure between the feet and the support surface to the central nervous system (Strzalkowski et al., 2018). This information is integrated with sensory information from the visual, vestibular, and proprioceptive systems to allow the CNS to provide the most appropriate motor responses to maintain postural equilibrium.

Four types of cutaneous mechanoreceptors can be found within the glabrous skin of the feet: slow adapting type 1 receptors, slow adapting type 2 receptors, fast adapting type 1 receptors and fast adapting type 2 receptors. Type 1 receptors have small receptive fields, while type 2 receptors have large receptive fields (Inglis et al., 1994; Strzalkowski et al., 2018). Although all cutaneous mechanoreceptors are readily activated in response to pressure and vibration, the slow adapting receptors (i.e., Merkel cells and Ruffini endings) are more sensitive and responsive to touch and pressure, while the fast-adapting receptors (Meissner and Pacinian corpuscles) are more sensitive to vibrations (Kennedy & Inglis, 2002; Strzalkowski et al., 2018). Thus, the fast-adapting receptors operate in an on-off response manner such that they are more responsive to the rate of change of mechanical stimuli while the slow adapting receptors operate in a sustained manner such that they fire throughout sustained stimuli. Additionally, the fast-adapting receptors are thought to signal contact velocity and slips that occur across the feet (Strzalkowski et al., 2018).

1.4.1. Distribution of the cutaneous mechanoreceptors across the glabrous skin of the feet

The different types of cutaneous mechanoreceptors are thought to be evenly distributed across the glabrous skin of the feet but a recent review article by Strzalkowski and colleagues (2018) noted some differences in the location of these mechanoreceptors. The toes appear to have a greater concentration of both slow adapting (30%) and fast adapting type 1 receptors (50%) while the lateral portion of the metatarsals have a greater concentration of fast adapting type 1 receptors (50%). In general, there are more receptors located in the lateral region of the foot soles (see Figure 2 of Strzalkowski et al., 2018). The distribution of these receptors across the feet could imply a functional significance as the toes and lateral region represent the anterior and lateral limits of the BOS, respectively. In the anterior-posterior direction, around 60% of the COM is located towards the toes and metatarsals during upright stance (Fernández-Seguín et al., 2014). In the medio-lateral direction, the lateral border makes more contact with the support surface during stance and gait. Therefore, the increased concentration of mechanoreceptors could reflect the importance of tactile feedback from these regions for postural control (Strzalkowski et al., 2018).

1.4.2. The role of the cutaneous mechanoreceptors for reactive balance control

In healthy adults, tactile feedback allows for the early detection of any change in foot pressure when postural equilibrium is disturbed. When the feet are perturbed, the mechanoreceptors within the glabrous skin of the feet sense the change in pressures, pressure distribution and ground reaction forces. Once this tactile information is sent to the CNS, the CNS sends the appropriate motor strategy by scaling the motor response according to the magnitude of the change in foot pressure (Inglis et al., 1994; Maki et al., 1999; Perry et al., 2000). This allows the COP to move ahead of the COM to restore postural equilibrium (Rietdyk et al., 1999).

The contribution of tactile perception for postural control has been explored by using methods that either increase or decrease the sensitivity of these cutaneous mechanoreceptors. For example, when tactile sensitivity was reduced with foot cooling, there was a short (3 ms) delay in electromyographic (EMG) onset latency and a 14-23% increase in soleus and medial gastrocnemius EMG amplitude (Ferguson et al., 2017). In contrast, a study by Germano et al. (2016) noted an approximately 50% decrease in soleus, medial gastrocnemius and medial fibularis EMG amplitudes in response to a surface translation (Germano et al., 2016). One reason that may explain the contrast between the studies is that the perturbation amplitude (2 cm) used in the study by Germano et al. (2016) was smaller compared to the 20 cm perturbation amplitude used by Ferguson et al. (2017). So, it is likely that participants in the Germano et al. (2016) study were able to anticipate the acceleration nature of the platform and as a result, could compensate by decreasing the magnitude of their EMG responses (Carpenter et al., 2005).

It is also notable that the changes reported in the Ferguson et al. (2017) and Germano et al. (2016) studies are smaller relative to the declines observed in older adults and adults with peripheral neuropathies. This may be because when tactile input is decreased in healthy young adults, other sensory inputs can compensate to supply information to the CNS on any changes in balance (Joseph Jilk et al., 2014). In contrast, older adults and individuals with peripheral neuropathies commonly experience other sensory deficits such as a reduced number of afferent fibres, vestibular system dysfunction and retinopathy, or other visual disorders that decrease the function of the visual system (D’Silva et al., 2016; Nardone & Schieppati, 2004). Therefore, the ability of older adults and those with peripheral neuropathies to compensate for a decrease in tactile sensation may be more limited and consequently, result in a greater change in postural responses compared to the application of a foot cooling protocol to the healthy young adult population.

In addition to changes in EMG activity, foot cooling leads to a shift in balance recovery strategy towards a hip (Meyer et al., 2004) or a multiple step response (Perry et al., 2000). The need to take more steps to recover balance could be the result of a reduced ability to sense heel contact during the initial step when responding to an unexpected surface translation. Furthermore, there is an increased COM excursion towards the posterior BOS limit before foot lift when individuals are perturbed in the backward direction. This illustrates an increased reliance on cutaneous mechanoreceptors at the heel for sensing posterior BOS limits (Perry et al., 2000).

In contrast to the studies reducing tactile sensitivity, increases in tactile feedback has been achieved by providing textured or indented insoles, and other interventions to facilitate plantar sensation. For example, in one of the earlier studies, Maki et al. (1999) applied polyethene tubing across the plantar-surface boundary (i.e., from the base of the metatarsals to the medial, lateral and heel region) to stimulate the cutaneous mechanoreceptors. This method of plantar facilitation reduced stepping responses when perturbed in the forward direction and reduced backward COP excursion in younger adults and older adults. This is the opposite to the effects observed when tactile contribution is reduced or impaired (Maki et al., 1999). Thus, it may be worth exploring if enhancing the sensitivity of the receptors will improve balance control in both young and older adults (Maki et al., 1999, 2008).

1.5. Vibrating Insoles: Mechanisms

One method that has been used to enhance the sensitivity of the cutaneous mechanoreceptors is the wearing of vibration insoles. Ideally, the insoles are created by placing two vibrating elements under the fore foot region and one element under the heel region of an insole (Hijmans et al., 2007). However, other studies have placed vibrating elements in the medial arch region (Lipsitz et al., 2015; Miranda, Hsu, Gravelle, et al., 2016a; Miranda, Hsu, Petersen, et al., 2016b) or directly in shoes or sandals (Lipsitz et al., 2015; Wei et al., 2012).

Vibration insoles are proposed to enhance tactile perception through two mechanisms. The first is stochastic resonance, where the addition of white noise boosts weak sensory signals (Cham et al., 2016; Priplata et al., 2003, 2006). Signal boosting is achievable because white noise contains a variety of frequencies and thus, when an optimal level of white noise is added, there is resonance which causes the weak input signals to be boosted and more readily perceived (see Figure 1 of van der Groen et al., 2018; White et al., 2019). This mechanism is counterintuitive to regular thinking as noise has previously been thought to disturb signal detection and system performance. Within the context of balance control, stochastic resonance allows the cutaneous mechanoreceptors located within the foot soles to perceive signals that were deemed too weak to be perceived during postural tasks. In contrast to the benefits of stochastic resonance, Priplata et al. (2006) suggested that vibration insoles work by causing a strain on the neuronal membrane of the cutaneous mechanoreceptors such that there are changes in ion permeability. Consequently, the membrane potential is brought closer to threshold which can allow the mechanoreceptors to be more sensitive to tactile information. In both cases, the increased signal perception will then allow for increased additional sensory input to the CNS and consequently, improved balance control.

Regardless of the mechanism, two important properties of vibrating insoles need to be considered when attempting to enhance tactile perception. First, although it is yet to be determined if and what frequencies of noise are optimal for enhancing tactile sensation, most studies have used white noise filtered to 100 Hz (Aboutorabi et al., 2018). There also seems to be no consensus on the type of noise optimal for enhancing tactile sensation and therefore, all types of noise may be beneficial for tactile sensation (Cham et al., 2016). With regards to the optimal level of vibrating amplitudes, studies have often used 90% of an individual's vibration perception threshold and have noted improvements in tactile sensitivity (Hijmans et al., 2007; Lipsitz et al., 2015; Priplata et al., 2003, 2006).

1.5.1. Vibrating insoles, tactile perception and postural control

1.5.1.1. Vibrating insoles and tactile perception

Based on previous work demonstrating the benefits of vibratory stimuli and tactile sensitivity in the fingertips (Collins et al., 1997; Liu et al., 2002), Khaodhiar et al. (2003) explored the effects of vibrating insoles on vibration perception thresholds and Semmes-Weinstein filament detection thresholds in the foot of adults with moderate to severe diabetic neuropathy. Khaodhiar et al. (2003) assessed tactile ability by measuring vibration perception thresholds using a biothesiometer at the toe and Semmes-Weinstein filament detection thresholds at the toe and plantar surface of the foot. These researchers noted improvements in tactile perception of approximately 20% at the metatarsals and heel, suggesting that the addition of noise to a vibratory stimulus improves one's ability to perceive vibrations and pressures against the foot (Khaodhiar et al., 2003).

1.5.1.2. Effects of vibrating insoles on postural stability during quiet standing

Improvements in tactile perception with vibratory stimulation may be one reason for the improvements in static balance performance. Several studies have measured postural stability while individuals stood quietly on the vibrating insoles. For most of these studies, the amplitude and frequency parameters of the vibrating insoles were determined for each participant prior to undergoing the experimental protocol. Since the amplitude for most studies was set to a maximum 90% of sensory threshold, participants were blinded to the experimental intervention (Costa et al., 2007; Dettmer et al., 2015; Hijmans et al., 2007; Lipsitz et al., 2015; Priplata et al., 2003; Priplata et al., 2006). Further, changes in postural stability have been most often assessed using COP measures in the anterior-posterior and medio-lateral directions over a 30 s to 1 min period (Dettmer et al., 2015; Hijmans, 2008; Wang & Yang, 2012; Wei et al., 2012). When COP was not monitored, postural stability was determined by analyzing sway parameters of single markers placed on the right shoulder (Priplata et al., 2003; Priplata et al., 2006).

The results from these studies indicate that standing on the vibrating insoles leads to an increase in postural stability. For example, Priplata et al. (2002, 2003, 2006) showed a decrease in postural sway parameters (e.g., mean radius, maximum radius, total area and range of AP and ML sway) with vibration. These studies also showed a decrease in random walk parameters that reflect the reduced tendency of the body to drift from an equilibrium point. Additionally, reductions in postural sway due to the vibratory insoles have been observed across different populations, including young adults (up to an 8% reduction), older adults (up to a 20% reduction), adults with peripheral neuropathies (up to a 15% reduction) and adults impacted by stroke (up to a 15% reduction) (Priplata et al., 2002; Priplata et al., 2003; Priplata et al., 2006). When adults with diabetic neuropathy performed a distraction task while having vision occluded during a quiet standing task, the vibrating insoles reduced COP velocity by 16% (Hijmans, 2008). Furthermore, when older adults stood with a sway-referenced visual surround, there was a 23% reduction in total COP sway length when standing on the insoles (Dettmer et al., 2015). In addition, Wang et al. (2012) analyzed the COP scaling exponent in the AP and ML directions following a 30 s quiet standing task with and without vibrating insoles. They found that insoles lowered the COP scaling (i.e., greater stability) component only in older adults and primarily in the AP direction. Lastly, Wei et al. (2012) and Costa et al. (2007) noted a 15-22% increase in COP complexity scores, reflecting greater COP variability, when older adults stood with vibrating insoles. Based on these studies, it is evident that vibrating insoles have the potential to reduce postural sway, through reductions in COP velocities, shoulder sway parameters and total COP sway length, during standing. Additionally, the increase in variability to the COP signal following improvement of tactile perception by the vibrating insoles may indicate an increased flexibility of the tactile system to respond to change in BOS conditions such as when an individual has to respond to a surface perturbation.

Vibratory insoles generally lead to greater improvements in balance performance in older compared to younger adults. This could be due to younger adults having a better tactile ability. However, it is important to note that improvements in balance control can be more commonly observed in young adults if visual input is completely restricted (Priplata et al., 2002; Priplata et al., 2003). For example, Dettmer et al. (2015) noted no changes in balance stability within younger adults when these individuals stood on a platform while the visual surrounding was sway referenced. In contrast, Priplata et al. (2003) noted improvements during a quiet stance task for both older and young adults when vision was completely restricted. Thus, although the vibrating insoles seem to be more beneficial for older adults or adults with decreased tactile sensation, there might be some added benefits of postural stability in specific contexts (i.e., visual restricted contexts) for younger adults. In particular, since there are studies that have noted greater effects of the insoles on postural stability under conditions where there is decreased visual contribution such as in eyes closed conditions or visual conflict conditions, this suggests that the insoles may have benefits for more complex conditions or in situations when there is increased reliance on the tactile system (Dettmer et al., 2015; Hijmans et al., 2008).

1.5.2. Vibration and Reactive Balance Control

To date, only one study has examined how vibration to the foot soles affects balance in response to an unexpected perturbation. Dettmer et al. (2016) examined whether young and older adults wearing vibrating insoles would respond differently to being perturbed in the backward direction. No changes in corrective postural responses, as noted by the lack of differences in force response latencies, maximum COP excursion values and total AP displacement, were observed when standing with or without tactile noise stimulation. However, several methodological limitations may have caused this lack of effect. First, the older adult participants likely already had good balance control since they were able to maintain their balance as well as young adults (Dettmer et al., 2016). Thus, the older adults may have been able to compensate for the experienced loss of balance by relying on their other sensory systems. Second, participants were perturbed while their eyes were open. When loss of balance is experienced while visual input is allowed, visual input provides an external frame of reference and information on vertical orientation (Horak, 2006; Horak et al., 1997). With this information, the CNS is better able to provide the appropriate motor response (Joseph Jilk et al. (2014). Third, the study by Dettmer et al. (2016) used unidirectional (backward) translations and participants were given a go signal before each surface translation. Thus, it is possible that the participants were familiar with and could anticipate the direction of the surface translation for each trial. This could have minimized the effects of the vibration since postural responses can be initiated earlier when individuals can anticipate the perturbation magnitude and direction of an upcoming trial (Horak et al., 1989).

1.6. Rationale, Purposes and Hypotheses

1.6.1. Rationale

Previous studies have demonstrated the potential of vibratory noise input to the foot soles to improve standing balance control across different populations (e.g., young adults, older adults, individuals with diabetic neuropathy, individuals impacted by stroke, etc.). However, there is little research on whether the benefits of vibration extend to reactive balance control and an individual's ability to generate corrective postural responses. This is important because most falls occur during a dynamic situation, where a trip or slip is experienced. If applying vibration to the foot soles is shown to be beneficial for reactive balance control, this might prove to be a useful balance aid for daily activities and potentially reduce the risk for falls.

The effects of vibratory noise input on postural control have primarily been limited to center of pressure (COP) measures for static stance research. While it is essential to explore changes in COP measures following improved tactile perception, there is a need to explore changes in muscle responses and kinematics to offer more understanding to the contribution of the tactile system toward balance control. Understanding the relationship between muscle responses, kinematics, COP and the tactile system can offer more explanation to the role of the tactile system during balance recovery.

1.6.2. Purpose and Hypotheses

The primary purpose of this thesis was to examine whether vibratory noise input applied to the plantar surface of the feet affects postural responses to an unexpected support surface translation. It was hypothesized that when vibratory noise input is applied to the foot sole, there would be earlier EMG onset latencies of the ankle muscles, decreased ankle joint angles and a smaller displacement of the body's COM. These altered postural responses to the perturbation were expected to occur due to an increase in tactile perception caused by the vibratory noise input and would allow for improved balance performance.

The secondary purpose of this thesis was to replicate previous findings by examine whether vibratory noise input affects static balance performance measures. It was hypothesized that there would be a smaller displacement of the body's COM and a smaller displacement of the COP when vibratory noise input was applied to the feet.

2. Methods

2.1. Participants

Eighteen adults (10 F, 8 M), with a mean ± 1 standard deviation (SD) age of 22.9 ± 3.2 y, mass of 70.6 ± 12.1 kg and height of 1.70 ± 0.10 m, participated in this study. Participants were excluded from this study if they self-reported any neurological, musculoskeletal, orthopedic, cardiovascular, or other balance-affecting disorders. Lastly, it was ensured that participants could feel the vibration of the tactors (see section 2.4) when at its highest setting so that sensory thresholds could be determined. Participants were also asked to refrain from exercising 48 hours before the experimental session.

Prior to the experimental session, all participants provided written informed consent. All procedures were approved by the Brock University Research Ethics Board.

2.2. Experimental Set-Up

Each participant visited the lab for a single experimental session. First, anthropometric measurements, including height, weight, leg length, knee width, ankle width, elbow width, wrist width and humeral offset were taken from each participant. These values were used for subject kinematic calibration in Vicon Nexus software (Vicon Nexus v 2.12, Vicon Motion Systems Ltd., Oxford, UK).

Next, the sensory threshold of each vibrating tactor location was determined while the participant was in a standing position on the force platform (OR6-2000, AMTI, Watertown, MA, USA). Identical to the quiet standing trials, participants stood with their eyes closed and wore headphones/earplugs to restrict visual and auditory sensory inputs when sensory thresholds were determined. Sensory threshold was determined at each of the six sites (i.e., the first metatarsal, fifth metatarsal and heel of each foot) and was defined as the smallest vibration amplitude from the vibrating tactors that was first perceived by the participant. This was achieved by increasing the vibration amplitude of each tactor from the lowest gain level until the participants first reported feeling the vibration. This was repeated two more times to confirm the sensory threshold. Once the sensory threshold was determined for each tactor location, a sub-sensory amplitude corresponding to approximately 90% of the sensory threshold was calculated. This level of vibration was then tested twice while standing to ensure that the participants could not feel the vibration. Since sensory thresholds are subject to change due to various reasons (e.g., skin deformation from standing for prolonged periods of time), sensory thresholds were re-tested a total of five times throughout the experimental protocol (Smith et al., 2022). The threshold tests occurred at the start of the protocol, prior to the standing trials protocol, prior to completion of the surface translation trials, midway through the surface translation trials and at the end of the experimental protocol.

Once the initial sensory thresholds were determined, participants were asked to lay on a massage bench so that pairs of surface electromyography (EMG) electrodes (Ag–AgCl, Covidien, Mansfield, MA, USA) with an interelectrode distance of 1 cm were placed over the soleus and tibialis anterior of both legs. A single ground electrode was placed over the right tibial tuberosity. Prior to all electrode placement, skin sites were prepared through shaving and cleaning of the skin with alcohol and a mild abrasive gel (Nuprep, Weaver and Company, Aurora, CO, USA). All EMG signals were amplified 350 times (Motion Lab Systems, Baton Rouge, LA, USA) and recorded using a data acquisition program (Spike 2, Cambridge Electronics Design, Cambridge, UK) at a sampling rate of 2000 Hz (micro1401, Cambridge Electronics Design, Cambridge, UK). Following electrode placement, participants were fitted into a safety harness that was later attached, via a rope, to an overhead track.

Next, each participant performed two maximum voluntary contractions (MVCs) for each muscle of interest. All contractions were performed while the participant was seated. During all MVCs, resistance was applied by the experimenter and each participant was instructed to contract as hard as they can for 3 s. For the tibialis anterior, the experimenter provided resistance by placing their hands over the participant's toes, while the participant slowly dorsi-flexed until maximum effort. For the soleus, the experimenter provided resistance by placing their hands over the participant's knees while the knees were at approximately 90 degrees. Then the participant slowly plantar flexed, with their toes planted on the ground, until maximum effort. Each MVC trial was separated by approximately 1 min to minimize fatigue.

After completion of the MVCs, 39 spherical retroreflective markers were placed on different bony markers, including four markers on the head (one marker each on the left and right temple and one marker each on the left and right back head in line with the frontal markers), four markers on the torso (one marker each on the C7, suprasternal notch, the T10 and the xiphoid process of the sternum), 14 markers along both arms (one marker each on the left and right acromio-clavicular joint, upper lateral one-third surface of the arm, lateral epicondyle, lower one-third lateral surface of the forearm, radial styloid process, ulnar styloid process and third metacarpal), one marker on the right scapula, 16 markers along each leg (one marker each on the left and right anterior superior iliac spine, posterior superior iliac spine, lower lateral one-third surface of the thigh, knee joint line, lower one-third lateral surface of the shank, lateral malleolus, calcaneus and second metatarsal head) using double sided tape. From these markers, the Vicon Plug-in-Gait Full body model was formed within the Vicon Nexus software (Vicon Nexus v 2.12, Vicon Motion Systems Ltd., Oxford, UK). Additionally, one marker was placed on each corner of the force platform to track its location. The position of all markers was monitored using an eight-camera three-dimensional motion capture system (Vicon Motion Systems Ltd., Oxford, UK). All kinematic data was sampled at 100 Hz.

2.3. Experimental Protocol

Following experimental set-up, participants performed six quiet standing trials followed by 28 surface translation trials. The quiet standing and surface translation trials were separated by a 5 min break to minimize any effects of fatigue on postural responses. Participants were also given a 2 min break halfway through completion of the surface translation trials and were told that additional breaks would be given if needed.

The quiet standing trials required participants to stand on top of a pair of rubber pads with tactors embedded (see section 2.4) placed on the force platform (OR6-2000, AMTI, Watertown, MA, USA). The force platform was locked in place and secured to a linear positioning stage (DRS-120-09-176-01, H2 W Technologies Inc., Valencia, CA, USA). A 1.8 m x 0.9 m wooden platform surrounded the force platform. All force plate data were recorded using a data acquisition program (Spike 2, Cambridge Electronics Design, Cambridge, UK) at a sampling rate of 1000 Hz (micro1401, Cambridge Electronics Design, Cambridge, UK). Participants stood on this force platform with a self-selected stance width. This self-selected stance width was kept constant across all quiet standing and surface translation trials by taping the pads to the force plate once the participant's self-selected stance width was determined.

For each quiet standing trial, participants stood quietly for 60 s, with their arms at their side and their eyes closed. Prior to each trial, the experimenter verbally reminded the participants to keep their eyes closed throughout the trial. Visual feedback was removed to increase reliance on the tactile system for balance control (Horak, 2006; Horak et al., 1997). Participants also wore headphones and earplugs during each trial to prevent them from hearing the sound of the vibrating tactors. The vibrating tactors within the rubber pads were turned on for three of the six standing trials and turned off for the remaining three trials (i.e., control trials). The vibration and control trials were presented in one of four counterbalanced orders for each participant. During each quiet standing trial, kinematic, kinetic and EMG data were collected.

For the 28 surface translation trials, participants stood on the same force platform as the quiet standing trials, with their arms at their side and their eyes closed. At a time unknown to the participant, the platform on which the participant was standing, displaced 0.35 m in either the forward or backward direction. Each translation consisted of a peak acceleration of 1 m/s^2 for a 400 ms period, a peak velocity of 0.3 m/s for an 800 ms period and a peak deceleration of 1 m/s^2 over a 400 ms period . Half of the 28 surface translation trials occurred in the anterior direction and the other fourteen were in the posterior direction. For half of the surface translation trials, the tactors began vibrating approximately 2-4 s prior to translation onset and remained active until approximately 2 s after the end of the translation. The order of direction and vibration trials were presented in one of four counterbalanced orders for each participant to reduce familiarization and anticipation of the direction and vibration condition. Additionally, prior to the completion of the surface translation trials, participants were shown a video of an individual on the moving platform to reduce familiarization and any anxiety that may arise from being on the novel moving platform.

In response to each surface translation, participants were instructed to recover their balance without taking a step. There was an intertrial interval of 5-10 s to ensure that participants were unaware of the timing of the next surface translation.

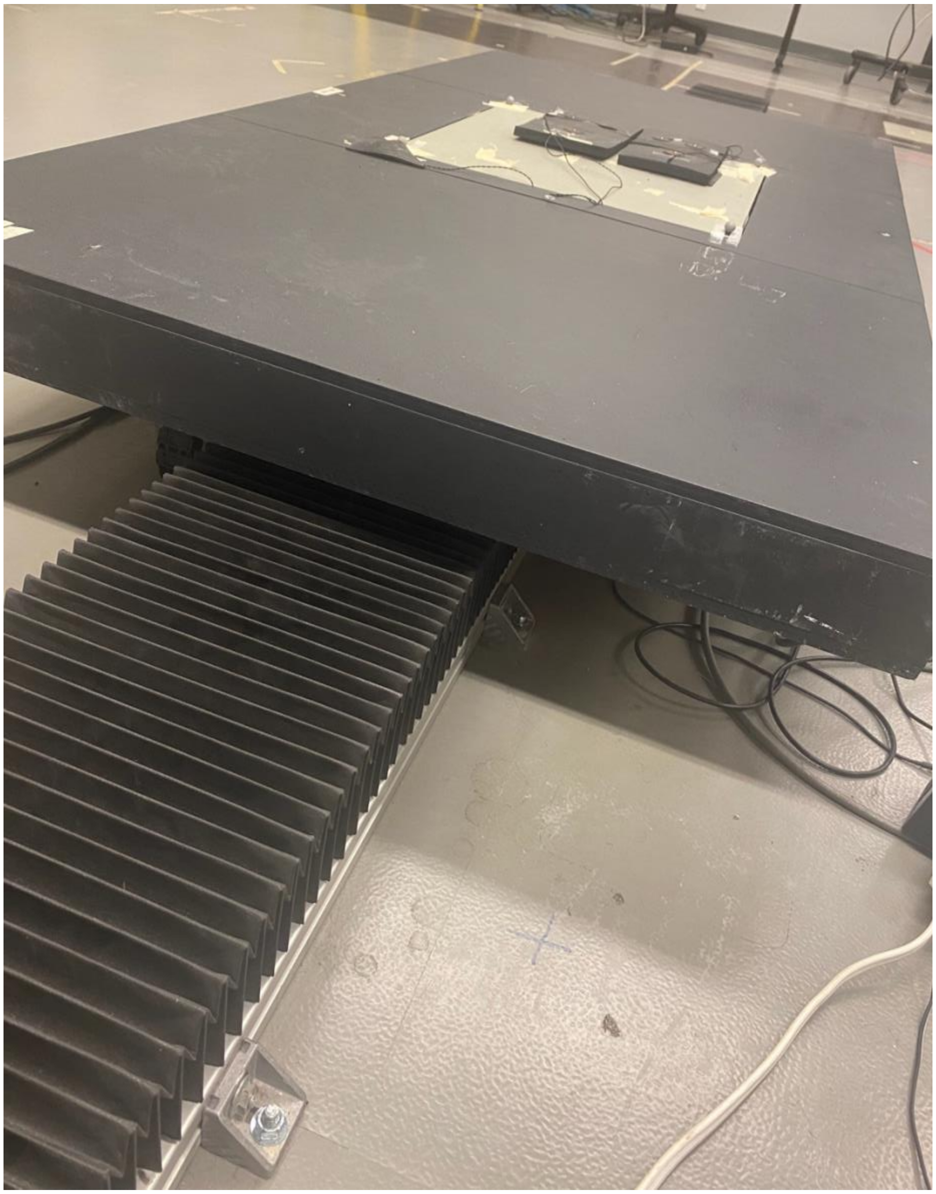
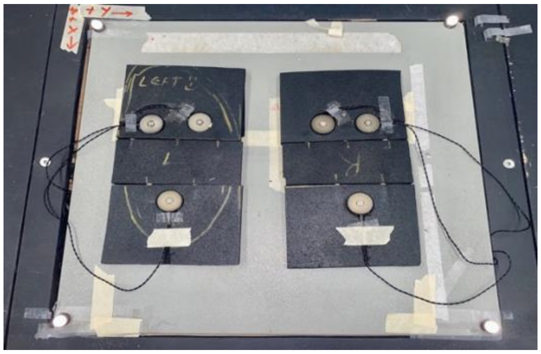


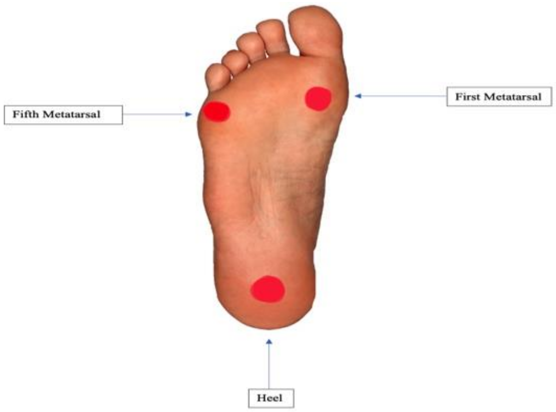
Figure 1: The surface translation platform used in this experiment. Note the force plate that was embedded in the middle of the wooden surround and the rubber pads, containing the vibrating tactors, were placed on top of the force plate.

2.4. Vibrating Tactors

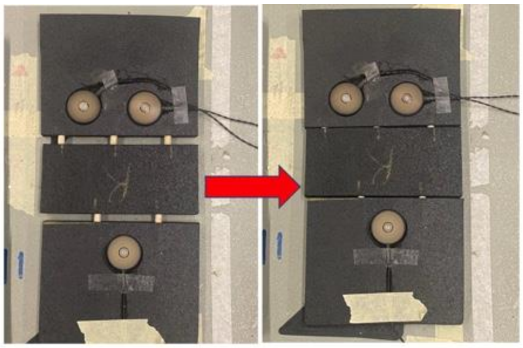
Participants stood with each foot on a separate rubber pad. Each pad consisted of three vibrating tactors (C-2; Engineering Acoustics, Winter Park, FL) imbedded within a rubber pad (Figure 2). The rubber pad was 0.5 inches thick while each tactor was approximately 3 cm wide and 0.79 cm tall. One tactor was embedded at each of the heel, fifth metatarsal region and the first metatarsal region of each pad so that the tactors vibrated at these specific regions (Figure 2A). Each tactor was embedded by drilling holes into the pad. The pads were made so that they were adjustable to account for participants with different shoe sizes (Figure 2C). The same vibration frequency was passed through all tactors but the vibration amplitude differed for each foot and tactor location. All six tactors were connected to a control box (Universal Controller; Engineering Acoustics, Winter Park, FL) and commercially available software (Tactor Development Kit (TDK); Engineering Acoustics, Winter Park, FL) was used to generate and adjust the amplitude of the vibration noise signals.



A.



B.



C.

Figure 2: A. The vibration device used in this experiment with tactor placement at the first metatarsal, fifth metatarsal and heel regions of both feet. B. Regions of the foot stimulated by the tactors on one rubber pad. C. The dowels, smaller pads and tactors used to make the vibration device. Different sized pads were attached using the dowels to account for different shoe/foot sizes.

2.5. Data Analysis

2.5.1. Kinematics

All kinematic data were collected using Vicon Nexus software (Vicon Nexus 2.11, Vicon Systems Ltd., Oxford, UK). Raw kinematic data was low-pass filtered at 5 Hz using a 2nd order Butterworth filter (Vicon Nexus 2.11, Vicon Systems Ltd., Oxford, UK) (Rajachandrakumar et al., 2018). Whole body COM position was calculated within Vicon Nexus based on the weighted-average position of 16 segments (pelvis, head, right and left femur, tibia, foot, humerus, radius, hand and thorax) over the duration of each trial and were later referenced to the position of the force platform. Relative knee joint angles were calculated as the angle between the relative orientations of the femur and tibia segments while relative ankle joint angles were calculated as the angle between the relative orientations of the tibia and foot segments. The 3D coordinates of the whole body COM position and joint angles were then exported as model outputs. Analyses of all kinematic data were limited to the sagittal plane.

For the standing trials, three measures including the peak-to-peak COM displacement, SD of COM displacement, and mean COM velocity were calculated over the entire 60 s trial (MATLAB and Simulink R2022a, The MathWorks, Inc., Natick, Massachusetts, US). The COM displacement was calculated as the average displacement of COM position values from the mean COM position . From the COM displacement, the peak-to-peak displacement, SD of the displacement and the mean velocity of the COM was calculated. The peak-to-peak COM displacement indicated the range of excursion of the COM, the SD of the COM displacement indicated the variability of the COM, while the COM velocity determined the rate of movement of the COM during each quiet standing trial.

For the surface translation trials, six measures, including the COM displacement, left and right knee and ankle angles, were calculated (MATLAB and Simulink R2022a, The MathWorks, Inc., Natick, Massachusetts, US). COM displacement was referenced to the position of the platform at each instant in time over the duration of each trial. COM displacement, left and right knee and ankle angles values were recorded during 0 ms, 100 ms, 200 ms, 300 ms and 400 ms post translation onset. Kinematic data was analyzed up to 400 ms post translation onset to determine responses that corresponded to the acceleration interval of the platform. Due to errors in kinematic modeling, data from one participant could not be analyzed.

Stepping responses were also recorded for each participant. A step was recorded if the second metatarsal moved at least 10 mm vertically and 5 mm in the AP direction, and the calcaneus displaced at least 5 mm in the AP direction. All recorded stepping responses were visually confirmed by the experimenter in Vicon Nexus.

2.5.2 Muscle Activity

All EMG signals were full-wave rectified and band pass filtered between 30-500 Hz with a 4th order Butterworth filter (Spike2, Cambridge Electronics Design, Cambridge, UK) (McIlroy & Maki, 1993c). For each standing trial, the EMG amplitude of each muscle was determined as the root mean square (RMS) amplitude over the entire 60 s trial. To facilitate comparison between participants, all EMG amplitudes were normalized to each participant's MVC amplitudes. MVC amplitudes for each participant were achieved by calculating the maximal RMS amplitude, within a 200 ms window, observed during each muscle's MVC trials. The average maximum RMS amplitude from the two MVC trials was set as 100% MVC.

For each surface translation trial, EMG activity of the TA and SOL for the right and left legs were quantified in two ways: the EMG onset latency and the EMG amplitude. The EMG onset latency for each muscle was determined as the time when the EMG signal first exceeded one SD from the mean baseline for a minimum of 25 ms after translation onset (Tokuno et al., 2006) (Figure 3). Baseline EMG activity was considered to occur during the 200 ms period prior to translation onset. All EMG onset latencies were later visually confirmed by the experimenter. The EMG amplitude was calculated as the RMS amplitude of the band-pass and rectified EMG signal during three time intervals: 200 ms before surface translation to represent background EMG activity, 0-200 ms after muscle onset and 200-400 ms post muscle onset (Ferguson et al., 2017). Background EMG activity was examined to determine whether there was anticipatory activity and whether this differed between vibration conditions. The two time intervals post muscle onset were examined to reflect the early (reflexive) and late (volitional) responses. For muscle onsets that could not be confirmed, EMG amplitudes were calculated for the 200 ms before surface translation, 0-200 ms after translation onset and 200-400 ms post translation onset.

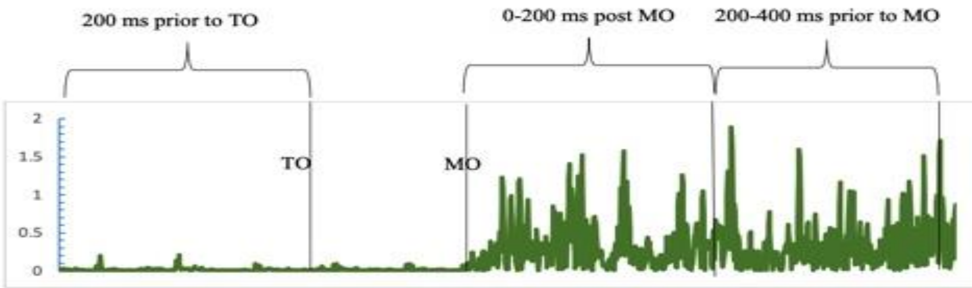


Figure 3: An example trace of EMG onset time and EMG amplitudes during the three time conditions (i.e., 200 ms prior to translation onset, 200 ms post muscle onset and 400 ms post muscle onset). TO = Translation Onset, MO= Muscle Onset

2.5.3. Centre of Pressure (COP)

Force plate data were low pass filtered at 5 Hz using a 4th order Butterworth filter (Spike2, Cambridge Electronics Design, Cambridge, UK) (Ferguson et al., 2017). From the filtered ground reaction force and moment signals, centre of pressure (COP) position was calculated in the AP direction using the equation $COP_{ap} = (-Mx)/Fz$ where Mx refers to the moment in the AP direction and Fz refers to the vertical force. For the standing trials, three COP measures were analyzed: the peak-to-peak amplitude, standard deviation and mean velocity. The COP peak-to-peak amplitude was derived as the difference between the maximum and minimum position within each 60 s trial and was used to reflect the range of COP excursion. The standard deviation of the COP was derived as the average dispersion of COP position from the mean COP position over each 60 s trial and was used to indicate the magnitude of COP variability during each trial. Lastly, the COP mean velocity was calculated as the mean of absolute COP velocity values and was used to determine the rate of movement of the COP. COP measures were not examined for the surface translation trials.

2.6. Statistical Analyses

To determine whether vibratory noise input affected quiet standing, separate paired t-tests (standing with and without vibration) were conducted for each kinematic, kinetic and EMG dependent variable. In contrast, the effect of vibration on postural responses to an unexpected perturbation was assessed in two ways. First, the EMG onset latencies between the no vibration and vibration trials were compared by performing separate paired t-tests for each muscle. The first four surface translation trials for each participant were excluded from statistical analyses to reduce any potential first trial effects. Further, the average EMG onset latency for a given participant was only included in the statistical analysis if an EMG latency was observed for at least four trials within a particular translation direction. Second, since assumptions of normality were not met for the EMG amplitudes and the kinematic measures, the non-parametric Friedman test was performed for these measures. For the EMG amplitudes, the Friedman test equivalent of a 2 (with vs. without vibration) x 3 (200 ms before translation onset vs. 0-200 ms post- muscle onset vs. 200-400 ms post-muscle onset) repeated measures ANOVA was conducted. For the kinematic measures, the Friedman test equivalent of a 2 (with vs. without vibration) x 5 (translation onset vs. 100 ms post translation onset vs. 200 ms post translation onset vs. 300 ms post translation onset vs. 400 ms post translation onset vs. 500 ms post translation onset) repeated measures ANOVA was performed. Similar to the EMG onset latencies, the first four surface translation trials for each participant were excluded from statistical analyses to reduce any potential first trial effects.

For all measures, outliers were deemed as values that were at least three standard deviations from the mean but were included in the statistical analyses to examine whether participants could potentially be classified as responders or non-responders to the vibratory noise input. Further, data from the surface translation trials were statistically analyzed twice: once, when data included the stepping trials and another when the stepping trials were excluded. This was performed to determine whether the observed stepping responses influenced the statistical outputs of the kinematic and EMG measures.

All statistical analyses were conducted using IBM SPSS Statistic (version 24, Armonk, NY, USA). A significance level of $p \leq 0.05$ was used for all analyses.

3.1. Quiet Standing

3.1.1. Kinematics

Vibration applied during quiet standing did not affect any of the observed COM measures. No differences in peak-to-peak COM displacement ($t_{16} = -0.001$; $p = 0.999$), SD of the mean displacement ($t_{16} = -0.934$; $p = 0.364$), and the mean velocity of the COM ($t_{16} = -0.767$; $p = 0.454$) were found between the two vibration conditions (Table 1).

Table 1: Kinematic, kinetic and EMG measures obtained without and with vibration during the 60 s quiet standing trials. All values are reported as the mean ± 1 SD.

Measure	No vibration	Vibration
Peak-to-peak COM displacement (mm)	30.62 ± 9.77	30.62 ± 12.67
SD of COM displacement (mm)	6.40 ± 2.29	6.73 ± 2.91
Mean COM velocity (mm/s)	4.08 ± 1.41	4.28 ± 1.87
Right TA (%MVC)	3.08 ± 1.44%	3.23± 1.71%
Right Soleus (% MVC)	7.99 ± 2.87%	8.24± 2.85%
Peak-to-peak COP displacement (mm)	45.10 ± 13.22	45.21 ± 17.05
SD of COP displacement (mm)	8.11 ± 2.64	8.29± 2.98
Mean COP velocity (mm/s)	11.61± 4.72	12.14± 7.8

3.1.2. Muscle Activity

When vibration was applied to the foot soles during standing, the right TA EMG activity was 3.23± 1.71% MVC while the right SOL EMG activity was 8.24± 2.85% MVC during the quiet standing period. These magnitudes of EMG activity were not different from when no vibration was applied as the right TA and SOL EMG activities were 3.08 ± 1.44% MVC (t_{17} = -1.658; p = 0.116) and 7.99 ± 2.87% MVC (t_{17} = -1.210; p = 0.243), respectively (Table 1). Since muscles of the left leg behaved similarly to the right leg and no significant differences were observed between conditions, a summary of results obtained from the left TA and SOL EMG activity are presented in Appendix 7.1.

3.1.3. Center of Pressure (COP)

The COP measures obtained during the quiet standing trials were not affected by vibration (Table 1). No differences were found in the peak-to-peak COP displacement (t_{16} = -0.04; p = 0.969), the SD of COP displacement (t_{16} = -0.37; p = 0.717) or the mean COP velocity (t_{16} = -0.546; p = 0.593) between the two vibration conditions.

3.2. Surface Translation

3.2.1. Kinematics: Stepping responses

During the no vibration condition, when vibration was not applied to the plantar surface of the foot sole, 11 of 18 participants stepped at least once in response to a forward surface translation. A stepping response was observed in 31 trials (out of a possible 102 trials) during this condition. This frequency of stepping responses was similar to when vibration was present, where ten participants stepped at least once, with a total of 32 stepping trials (out of a possible 102 trials) being observed.

Although fewer stepping trials were observed in response to a backward support surface translation, vibration did not appear to influence the frequency of stepping responses. Five stepping trials from three participants were recorded during the no vibration condition while in the vibration condition, four stepping trials from two participants occurred during the vibration condition.

To determine whether to include or exclude the stepping trials when statistically analyzing the kinematic and EMG measures, the data were analyzed with the stepping trials included and then with the stepping trials excluded. Since the statistical outcomes of the kinematic measures did not differ regardless of the statistical method (i.e., inclusion/exclusion of trials) and to maximize the number of trials and participants, it was decided to include all trials for further analysis. Thus, the following sections report the results when both stepping and non-stepping trials were included for analysis.

3.2.2. Kinematics: Centre of Mass

In response to a forward support surface translation, the COM moved (relative to the moving support surface) in the backward direction (Figure 4a). However, no differences in the magnitude of COM displacement were observed between vibration conditions at 0 ms ($Z = -0.213$, $p = 0.831$), 100 ms ($Z = -0.166$, $p = 0.868$), 200 ms ($Z = -0.071$, $p = 0.943$), 300 ms ($Z = -0.450$, $p = 0.653$) and 400 ms ($Z = -0.450$, $p = 0.653$) post-translation onset (Figure 4a). Vibration also had minimal influence on the amount of COM displacement when the support surface moved in the backward direction. No differences in forward COM displacement between vibration conditions were observed at 0 ms ($Z = -1.302$, $p = 0.193$), 100 ms ($Z = -1.917$, $p = 0.055$), 300 ms ($Z = -1.444$, $p = 0.149$) and 400 ms ($Z = -1.302$, $p = 0.193$) post-translation onset (Figure 4b). The lone difference occurred at 200 ms post-translation onset, when the COM was positioned in a more forward direction by 1 mm when vibration was applied to the foot sole ($Z = -2.059$, $p = 0.039$) (Figure 4b).

(a) (b)

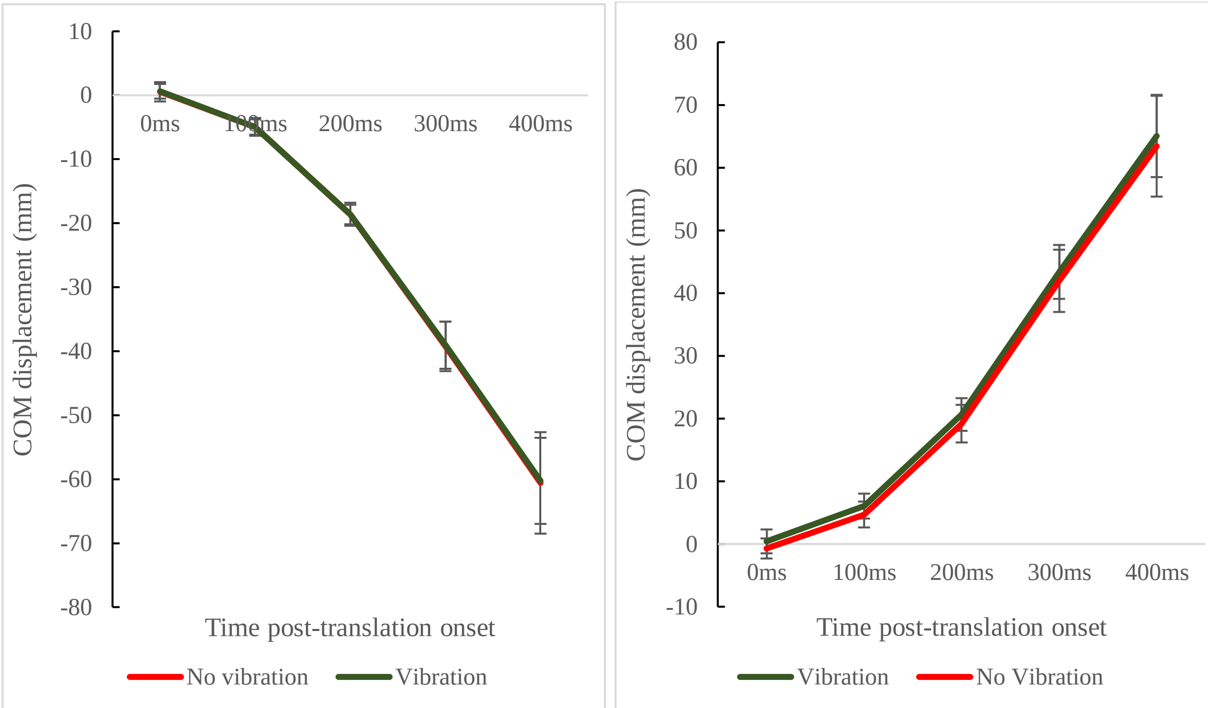


Figure 4: Mean ± 1 SD COM displacement at five timepoints post-translation onset, relative to the platform movement, in response to (a) a forward support surface translation and (b) a backward support surface translation during the two vibration conditions. A positive COM displacement indicates body movement in the forward direction.

<u>3.2.3. Kinematics: Joint Angles</u>

When the support surface moved in the forward direction, the right knee moved into a more flexed position from 0 ms to 400 ms post-translation onset. However, there were no differences in the right knee angle at 0 ms (Z =-1.538, p = 0.124), 100 ms (Z =-1.538, p = 0.124), 200 ms (Z =-1.538, p = 0.124), 300 ms (Z =-1.160, p = 0.246), and 400 ms post translation onset (Z =-1.065, p = 0.287) between vibration conditions (Figure 5a). Forward support surface translations also resulted in greater right ankle dorsi-flexion but no differences in right ankle angle were observed between the two vibration conditions at 0 ms (Z =-1.112, p = 0.266), 100 ms (Z =-1.207, p = 0.227), 200 ms (Z =-0.781, p = 0.435), 300 ms (Z =-0.639, p = 0.523), and 400 ms (Z =-0.308, p = 0.758) post-translation onset (Figure 6a).

Compared to when the support surface moved forward, there was generally less change in right knee and ankle joint angles in response to a backward translation. Analyses revealed no differences in right knee angle between vibration conditions at 0 ms (Z =-0.071, p = 0.943), 100 ms (Z =-0.166, p = 0.868), 200 ms (Z =-0.308, p = 0.758), 300 ms (Z =-0.734, p = 0.463), and 400 ms post-translation onset (Z =-1.207, p = 0.227) (Figure 5b). No differences in the right ankle angle were observed between vibration conditions at 0 ms (Z =-0.355, p = 0.723), 100 ms (Z =-0.213, p = 0.831), 200 ms (Z =-0.308, p = 0.758), 300 ms (Z =-1.160, p = 0.246), and 400 ms (Z =-1.681, p = 0.093) post translation onset (Figure 6b).

The left knee and ankle joint angles responded similarly to those in the right leg and therefore, a summary of the kinematic results for the left knee and ankle joint angles are presented in Appendices 7.2 and 7.3.

(a) (b)

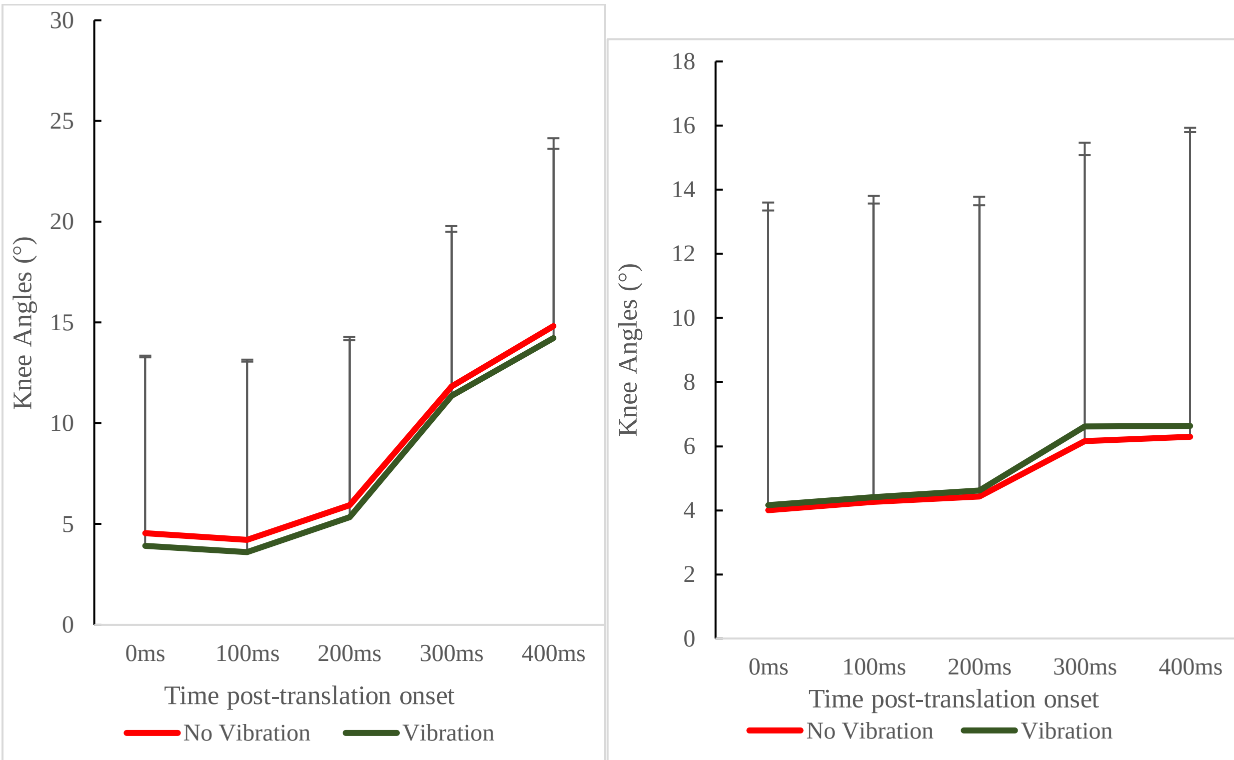


Figure 5: Mean + 1 SD knee angles at five timepoints post translation onset in response to a (a) forward and (b) backward translation for the two vibration conditions. 0° represents full extension while 90° represents full flexion

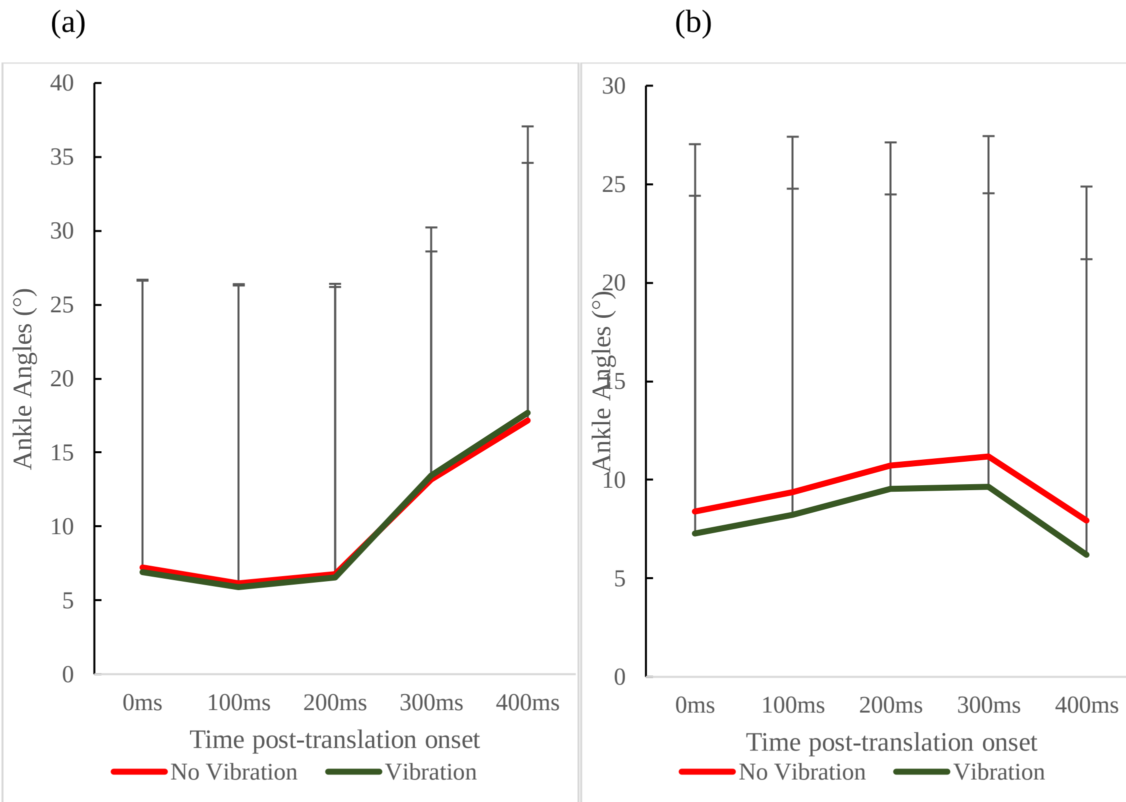


Figure 6: Mean + 1 SD ankle angles at five timepoints post-translation onset in response to a (a) forward surface translation and (b) backward surface translation. 0° represents full plantar-flexion while 90° represents full dorsi-flexion.

3.2.4. EMG Onsets

When participants experienced a forward support surface translation without vibration, the right TA and SOL were initiated 131.89± 7.76 ms and 172.85± 25.29 ms, respectively, after translation onset. These EMG onset latencies were not different when vibration was applied, as during this vibration condition the right TA and SOL EMG onset latencies were 130.87± 9.37 ms (t_{17} = 0.811; p = 0.429) and 166.33± 21.63 ms (t_{13} = 1.218; p = 0.245), respectively. Similarly, both the right SOL (t_{17} = 1.150; p = 0.266) and TA EMG onset latencies (t_{17} = 0.420; p = 0.680) were not influenced by vibration in response to a backward surface translation. Specifically, the right SOL was initiated 138.25± 24.87 ms while the TA was initiated later at 177.91± 38.93 ms post-translation onset in the no-vibration condition. When vibration was applied, the SOL was activated at 133.04± 14.14 ms while the TA was activated at 176.47± 30.97 ms post-translation onset. Since the muscles of the left leg behaved similarly to the right leg and no significant differences were observed in the two muscles of the left leg, a summary of results for the left TA and SOL EMG onset latencies are presented in Appendix 7.5.

3.2.5. EMG Amplitudes

Although the TA and SOL EMG activity increased from pre-translation (i.e., background EMG activity) to post-translation after a forward support surface translation, no differences in EMG amplitude were observed between the two vibration conditions at any given time interval. This was the case for the TA during the background EMG activity phase ($Z = -0.588$, $p = 0.557$), from 0-200 ms post-muscle onset ($Z = -1.154$, $p = 0.248$), and from 200-400 ms post-muscle onset ($Z = -0.631$, $p = 0.528$) as well as for the SOL (background EMG: $Z = -0.675$, $p = 0.500$; 0-200 ms post-muscle onset: $Z = -0.588$, $p = 0.557$; 200-400 ms post-muscle onset: $Z = -1.328$, $p = 0.184$) (Figures 7a and 7b).

In response to a backward support surface translation, there was more EMG activity in the SOL than the TA (Figures 7c and 7d). However, no differences were observed between vibration conditions in the SOL EMG activity during the 200 ms prior to translation onset ($Z = -1.459$, $p = 0.145$), as well as the 0-200 ms ($Z = -0.109$, $p = 0.913$), and 200-400 ms post-muscle onset intervals ($Z = -0.457$, $p = 0.647$). Similarly, the TA EMG activity was not different during the 0-200 ms ($Z = -0.109$, $p = 0.913$), and 200-400 ms post-muscle onset intervals ($Z = -0.457$, $p = 0.647$) (Figures 7c and 7d).

Since the muscles of the left leg behaved similarly to the right leg and no significant differences were observed in the left TA and SOL EMG amplitudes, a summary of results for these two muscles can be found in Appendix 7.4.

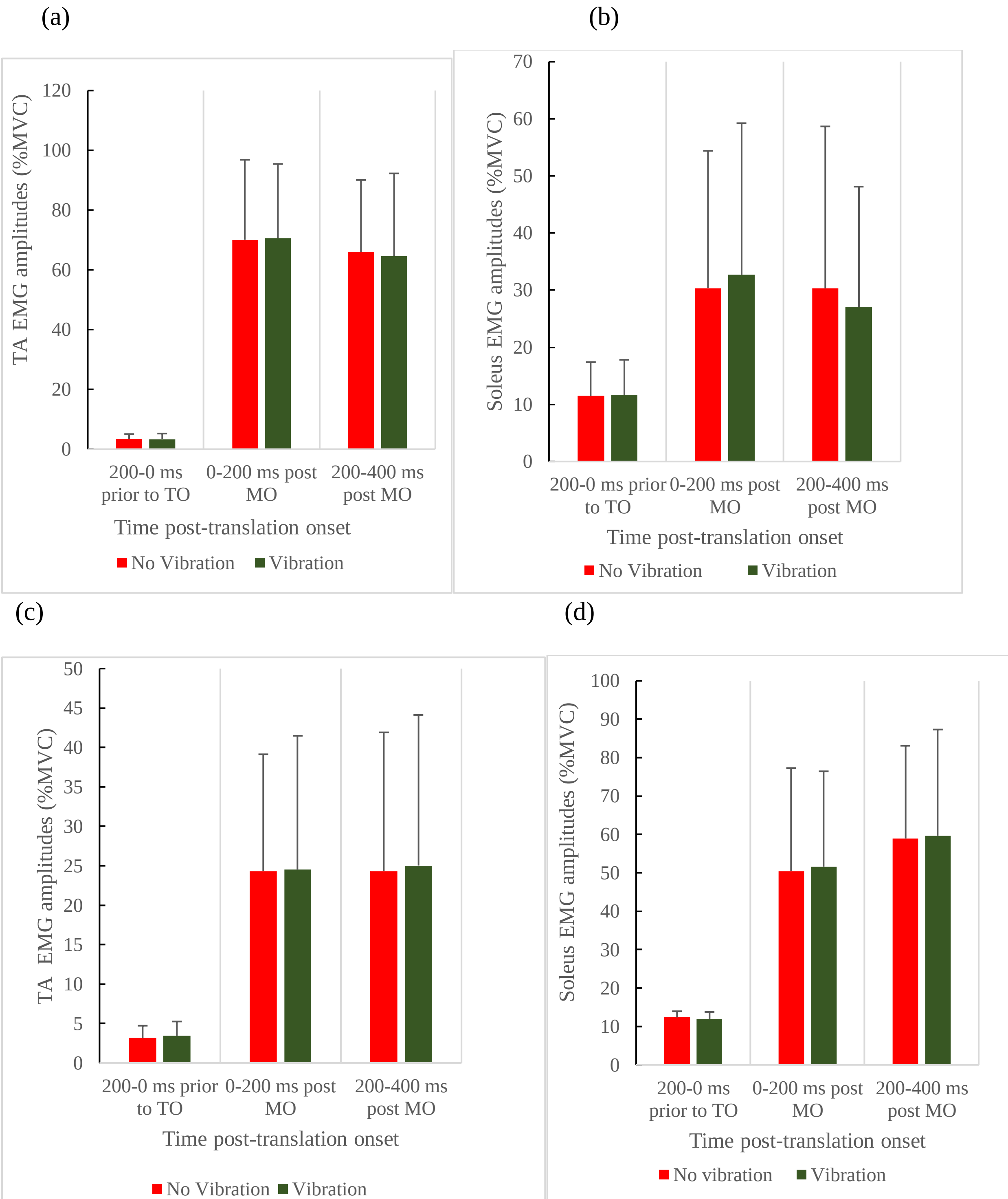


Figure 7: Mean + 1 SD EMG amplitude of (a) the tibialis anterior in response to a forward surface translation, (b) the soleus muscle in response to a forward surface translation, (c) the Tibialis Anterior muscle in response to a backward translation, and (d) the Soleus muscle in response to a backward translation between vibration conditions. TO= Translation Onset, MO= Muscle Onset.

4. Discussion

Previous research has shown that vibratory noise input applied to the plantar surface of the feet is beneficial for static balance control in older adults, adults with peripheral neuropathy and when vision is occluded, in younger adults (Priplata et al., 2002; Priplata et al., 2003, 2006). However, due to the limited research examining the effects of vibration on dynamic reactive balance control, this thesis aimed to determine whether vibration applied to the foot soles improves balance performance in younger adults in response to an unexpected perturbation. The effects of vibration on static balance performance were also examined to replicate previous findings. In contrast to the hypotheses, vibration applied to the foot soles did not influence static and reactive balance performance.

4.1. Static and Reactive balance performance

Based on previous research, it was hypothesized that vibration applied to the foot soles would improve static balance performance. However, this study found less than a 5% (i.e., statistically insignificant) difference in all COM, muscle activation and COP measures when standing without compared to with vibratory noise input. It was also hypothesized that with a more difficult task where one has to respond to an unexpected perturbation, there would be improvements in reactive balance performance. But in contrast to the hypothesis, this was not observed in this study. While vibratory noise resulted in a significant (1 mm) increase in COM displacement 200 ms after the onset of a backward translation, it is unlikely that this difference was large enough to be of functional relevance since all other joint angle, muscle onset latency and EMG activity measures were not different between the two vibration conditions. Specifically, there was less than a 5 °, 10 ms and 3% change in joint angles, EMG onset latencies and EMG amplitudes, respectively, between vibration conditions. The number of stepping trials was also not different between vibration conditions, which further supports the notion that the difference in COM displacement was unlikely to have had any meaningful change in dynamic balance control and consequently, the risk of falling.

The lack of change in static balance performance when vibration was applied is surprising since previous studies have reported that vibratory noise stimulation to the plantar surface of the feet improves static balance performance in younger adults (Priplata et al., 2002; Priplata et al., 2003). It was also surprising that vibratory noise stimulation to the foot soles had no influence on reactive balance performance given that the task (i.e., responding to an unexpected postural perturbation) is more difficult than simply quiet standing and participants were unable to rely on their visual or auditory feedback for balance control.

It is unlikely that the lack of vibratory effect on static and reactive balance could be attributed to participant characteristics. First, the magnitude of COM and COP sway and the magnitude of muscle activation observed in the younger adults in this study were similar to what has been reported in other studies. When standing with the eyes closed, previous studies have observed a 30-45 mm maximum COM and COP excursion, 8 mm in SD of COP, and 12 mm/s in mean COP velocity, all of which are similar to the results of this study (Chiari et al., 2000; Le Clair & Riach, 1996; Winter, 1995; Winter et al., 1998). Previous research has also noted more activity in the SOL compared to the TA during quiet standing. For example, Misiaszek and colleagues (2017) reported that TA muscle activity was 3.3 %MVC while SOL muscle activity was 13.3 %MVC during standing, both of which are similar to the 2-4 %MVC of TA EMG activity and the 7-11 %MVC of SOL activity found in this study (Misiaszek & Vander Meulen, 2017). The EMG onsets, EMG amplitudes, COM displacement and joint angles values observed in response to a platform perturbation in this study were also within range of previously reported values. For example, the TA has been shown to activate between 116-120 ms following the onset of a forward translation (Henry et al., 1998; Tokuno et al., 2006), while the SOL and gastrocnemius have been found to initiate approximately 105-110 ms following the onset of a backward translation (Henry et al., 1998; Tokuno et al., 2006). These values are comparable to this study's findings of 130-138 ms TA and SOL EMG onset latencies in response to a perturbation in the forward and backward directions, respectively. Similarly, previous studies have reported up to a 90-100 mm backward and forward COM displacement by 400 ms post-translation onset (Henry et al., 1998). These values are consistent with the results of this study, which noted a 70-80 mm COM displacement by 400 ms after perturbation onset. Since participants in this study stood and recovered their balance in a similar manner to individuals examined in previous studies, it can be assumed that the participants of this study were indeed healthy young adults and adopted typical balance control strategies. Further analyses of outlier data and comparisons of the variability of measures between vibration conditions suggested that it was unlikely that the participants consisted of a mix “responders” and “non-responders” that skewed the results.

Rather than individual differences, it is possible that the lack of results when vibratory noise input was applied to the foot sole could be attributed to methodological factors. While this study applied the vibration at similar foot sole locations (i.e., the first and fifth metatarsals, and the heel) and at similar vibration amplitudes (i.e., 90% sensory threshold) as previous studies, the vibratory noise input differed in terms of the vibration frequency. Unlike this study, which used non-filtered noise, previous studies that observed balance improvements with vibratory noise input in younger adults used 100 Hz band filtered noise (Priplata et al., 2002; Priplata et al., 2003). Although no studies have explored whether and what noise frequency is most beneficial for improving balance performance, it is known that fast adapting mechanoreceptors are more responsive to vibrations at higher frequencies (i.e., > 40Hz-250Hz) (Mildren et al., 2016). Thus, applying a vibratory stimulus within this frequency range might have yielded more significant changes in static and/or reactive balance.

It is important to acknowledge that sensory thresholds were retested at different times during the experimental protocol to account for potential changes in sensory thresholds due to skin deformation. Despite this, it was difficult to confirm that the amplitudes were actually at 90% of sensory threshold during each trial. Consequently, if the vibration was not truly delivered at 90% of threshold, this may have affected the possibility of eliciting improvements in balance performance as studies reporting balance improvements from vibratory noise input in younger adults have all used 90% sensory threshold as their amplitude level. One other study found that amplitude levels other than 90% of sensory threshold still reduced gait variability in older adults (Lipsitz et al., 2015), but it is not clear whether this is applicable for balance improvement in younger adults (Wells et al., 2005).

Another reason that could explain the lack of balance improvement via vibratory noise stimulation could be that the position of the vibratory elements used might not be best for tactile stimulation in younger adults. More often, the vibratory elements have been placed at the first metatarsal, fifth metatarsal and the heel of the foot, since these regions represent the anterior and posterior limits of the base of the support (Hijmans et al., 2007; Perry et al., 2000). However, the heel has been reported to have higher thresholds and a reduced number of cutaneous mechanoreceptors (Mildren et al., 2016; Strzalkowski et al., 2018) compared to other areas of the foot, such as the first and fifth metatarsals. Further, Maki et al. (1999) stimulated the medial and lateral borders and the first metatarsal, fifth metatarsal and heel via polyethene tubing and showed balance improvements in response to a perturbation. Therefore, it is possible that positioning additional vibratory elements at the heel or the lateral and medial borders of the foot could result in more significant balance improvements.

Lastly, it must be acknowledged that the chosen form of tactile stimulation might not be effective for improving dynamic reactive balance control in younger adults. Vibratory noise has been thought to improve tactile perception through a stochastic resonance phenomenon where subthreshold noise improves weak signals (Priplata et al., 2006). This would suggest that vibratory noise stimulation works better in populations with lowered tactile perception as opposed to younger adults with intact tactile sensation. Further, it is possible that providing vibrotactile feedback rather than enhancing tactile perception might be a more effective means of improving static and dynamic balance in younger adults. Previous studies for which balance improvements have been more commonly observed in younger adults have used textured insoles (Corbin et al., 2007), thicker plantar inserts (Foisy et al., 2015) or tactile cues (Rogers et al., 2001). Researchers have also shown reductions in maximum COM displacements and reaction time in response to a surface translation when young adults wore a vibrotactile feedback device based on plantar force measurement (Ma & Lee, 2017).

4.2. Directions for Future Research

Although there was a lack of vibratory noise input effect on static and reactive balance observed in this study, this thesis has addressed a gap in this research area. However, since this thesis could not replicate the previously reported improvements in static balance performance with vibratory noise input, it is difficult to ascertain whether this is because of problems with the model of vibration used in this study or whether this model of vibration is more suited for other populations. Thus, future research should first repeat this study in older adults or adults with significantly reduced tactile sensation to determine whether the vibration device used in this study has any benefits for static and reactive balance performance in these populations. It would also be helpful to systematically determine how different vibration parameters, such as noise frequency and vibration amplitude, affect balance control in younger adults as well as adults with decreased tactile sensation (e.g., older adults and adults with peripheral neuropathies).

5. Conclusion

This study aimed to determine whether vibratory noise input applied at the foot sole improves balance during quiet standing and in response to an unexpected perturbation. In contrast to the hypothesis, measures of balance control were largely unaffected by the presence of the vibratory noise stimulation. Since no clear conclusions can be made to explain the lack of change in balance control, it is important that future studies determine whether the method of tactile stimulation used in this study improves balance in populations with worsened balance ability before other future directions can be explored.

6. References

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7. Appendix: Summary of statistical results for dependent measures involving the left limb

7.1. Left Soleus and TA EMG activity means, SD and statistical results (Quiet Standing)

Measure	No vibration	Vibration	df	t value	Sig.
Left TA (%MVC)	3.95 ± 2.39%	3.96± 2.48%	17	-.281	.782
Left Soleus (% MVC)	7.59 ± 3.74%	7.45± 3.48%	17	.505	.620

7.2. Left Knee joint angles means, SD and statistical results

Translation direction	Time interval	Vibration Condition	N	Mean	SD	Z value	Sig.
Forward	0 ms (translation onset)	No vibration	17	1.76	7.15	-.071	0.943
		Vibration	17	1.70	7.23		
	100 ms post translation onset	No vibration	17	1.34	7.22	-.213	0.831
		Vibration	17	1.31	7.29		
	200 ms post translation onset	No vibration	17	3.10	7.08	-.497	0.619
		Vibration	17	3.13	7.11		
	300 ms post translation onset	No vibration	17	10.25	9.88	-.260	0.795
		Vibration	17	10.21	10.57		
	400 ms post translation onset	No vibration	17	12.41	8.18	-.923	0.356
		Vibration	17	12.01	8.59		
Backward	0 ms (translation onset)	No vibration	17	1.54	7.03	-.071	0.943
		Vibration	17	1.62	7.02		
	100 ms post translation onset	No vibration	17	1.83	7.01	-.166	0.868
		Vibration	17	1.88	7.01		
	200 ms post translation onset	No vibration	17	2.14	7.03	-.071	0.943
		Vibration	17	2.17	7.02		
	300 ms post translation onset	No vibration	17	4.15	7.37	-.308	0.758
		Vibration	17	4.10	7.34		
	400 ms post translation onset	No vibration	17	3.94	7.69	-.308	0.758
		Vibration	17	3.96	7.55		

7.3. Left Ankle joint angles means, SD and statistical results

Translation direction	Time interval	Vibration Condition	N	Mean	SD	Z value	Sig.
Forward	0 ms (translation onset)	No vibration	17	8.77	13.73	-1.112	0.266
		Vibration	17	1.70	7.23		
	100 ms post translation onset	No vibration	17	7.42	14.75	-1.207	0.227
		Vibration	17	4.50	12.59		
	200 ms post translation onset	No vibration	17	7.63	15.11	-1.065	0.287
		Vibration	17	4.47	14.55		
	300 ms post translation onset	No vibration	17	15.71	12.60	-.166	0.868
		Vibration	17	13.15	10.94		
	400 ms post translation onset	No vibration	17	23.30	19.07	-.686	0.492
		Vibration	17	20.62	14.33		
Backward	0 ms (translation onset)	No vibration	17	7.16	9.83	-1.538	0.124
		Vibration	17	3.54	18.56		
	100 ms post translation onset	No vibration	17	8.26	9.41	-1.538	0.124
		Vibration	17	4.81	16.70		
	200 ms post translation onset	No vibration	17	9.52	9.03	-1.728	0.084
		Vibration	17	5.86	15.92		
	300 ms post translation onset	No vibration	17	11.52	10.55	-1.586	0.113
		Vibration	17	6.96	13.54		
	400 ms post translation onset	No vibration	17	3.66	15.13	-1.302	0.193
		Vibration	17	0.82	23.65		

7.4. Left TA EMG Amplitudes (forward surface translation)

Time interval	Vibration Condition	N	Mean (% MVC)	SD	Z	Sig.
200 ms prior to TO	No vibration	18	4.23	2.49	-.588	.557
200 ms prior to TO	Vibration	18	4.16	2.58		
200 ms post MO	No vibration	18	79.86	30.28	-.849	.396
200 ms post MO	Vibration	18	84.55	48.07		
400 ms post MO	No vibration	18	79.45	41.41	-.152	.879
400 ms post MO	Vibration	18	81.43	49.62		

7.5. Left SOL EMG Amplitudes (backward surface translation)

Time interval	Vibration Condition	N	Mean (% MVC)	SD	Z value	Sig.
200 ms prior to TO	No vibration	17	11.26	5.11	-1.111	.267
200 ms prior to TO	Vibration	17	11.13	5.33		
200 ms post MO	No vibration	17	60.12	32.83	-1.198	.231
200 ms post MO	Vibration	17	58.04	30.66		
400 ms post MO	No vibration	17	65.47	38.18	-.240	.811
400 ms post MO	Vibration	17	65.80	45.44		

7.6. Left EMG Onset Latencies

Muscle	Translation direction	Vibration condition	N	Mean	SD	t	Sig.
TA	Forward	No vibration	18	131.20	8.52	.629	.538
		Vibration	18	130.36	9.86		
SOL	Backward	No vibration	18	139.13	19.45	1.185	.252
		Vibration	18	134.97	18.19		